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Sagittal plane biomechanics cannot injure the ACL during sidestep cutting

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1. Introduction

Anterior cruciate ligament (ACL) injury is a common and potentially disabling sports related injury. Approximately 80,000 ACL injuries occur annually within the United States, with roughly 50,000 requiring surgical reconstruction, at a cost of almost one billion dollars

(Daniel and Fritschy, 1994). Approximately 70% of ACL injuries occur as a result of a non-contact episode, typically during the execution of movements characterized by a sudden deceleration or direction change, such as sidestep cutting (Arendt and Dick, 1995; Griffin et al., 2000). Of particular concern, is the disproportionate incidence of non-contact ACL injuries based on gender, with females reported to suffer these injuries 5–7 times more frequently than males (Arendt and Dick, 1995). Despite the vast amount of ongoing research into ACL injuries, the precise mechanisms of non-contact

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ACL injury, and the extent to which they may be gender specific, remain unclear. Theories continue to evolve alongside this research however, as to the most likely contributors to ACL injury risk.

Sagittal plane mechanisms for non-contact ACL injury have been proposed previously for sports movements (Chappell et al., 2002; DeMorat et al., 2004; Griffin et al., 2000). Such postulates are based on the fact that the landing phase of these movements typically incorporates large quadriceps force at relatively small flexion angles, a combination known to induce anterior force on the tibia (Durselen et al., 1995; Pandey and Shelbourne, 1997). Women are often observed to perform these movements with less knee flexion than males (Chappell et al., 2002; Malinzak et al., 2001), which is thus viewed as a likely contributor to their increased risk of ACL injury (Colby et al., 2000; Griffin et al., 2000; Lephart et al., 2002). The neuromuscular control and strength ratio of the hamstrings and quadriceps are also viewed as important components of a sagittal plane injury mechanism (Colby et al., 2000; Griffin et al., 2000). Both of these variables have similarly been found to differ across gender (Wojtys et al., 2003).

Another important component of the sagittal plane loading mechanism during execution of sports movements is the presence of a large ground reaction force (GRF), which is directed posteriorly with respect to the tibial axis (McLean et al., 2004a). This force would help protect the ACL during the landing phase of these movements, but has not been taken into account in current theories on sagittal plane contributions to ACL injury. Thus, the potential for sagittal plane biomechanics to induce ACL injury may be overestimated.

If the sagittal plane biomechanics associated with sporting postures can produce an ACL injury, then prevention strategies could focus on teaching women to perform movements with more knee flexion, and more hamstrings activation. However, the true potential for ACL injury via this mechanism remains unclear, as ligament forces have not been measured or estimated during an injury-causing event. Furthermore, the need to examine the knee joint loading response to controlled systematic movement variations, or to evaluate injury scenarios, makes elucidation via human experimentation unfeasible. The recent development and validation of subject-specific forward dynamic simulations of sporting postures such as sidestep cutting, has made it possible to predict the effect of perturbations in neuromuscular control on resultant knee movement and loading (McLean et al., 2003). Models of this type provide a fast and relatively inexpensive means to study acute knee joint injuries while controlling all aspects of neuromuscular control (NMC). Using such an approach, the current study determined the effects of random variations in NMC during the stance phase of sidestep cutting on 3D knee loading. From these data, the

potential for the sagittal plane loading mechanism, comprising quadriceps and hamstring forces, flexion angle, and external anterior-posterior joint loads, to produce ACL injuries during sidestep cutting was evaluated and compared across gender.

2. Methods

Twenty subject-specific forward dynamics models of the stance phase (0–200 ms) of a sidestep cut were generated for the current study. Subject data implemented within each model were obtained from 10 male and 10 female NCAA Division I basketball players, whom were matched for experience level (Table 1). Prior to experimentation, approval for the research was gained through the Institutional Review Board of the Cleveland Clinic Foundation and written informed consent for all subjects was obtained. Subject inclusion in the study was based on no history of operable lower limb joint injury. A summary of subject characteristics is presented in Table 1.

2.1. Data collection

Three-dimensional (3D) kinematic and GRF data were recorded for each subject across 10 sidestep cutting trials. Approach speeds were monitored and required to fall between 4.5 and 5.5 ms⁻¹, reflecting speeds at which these movements are typically executed in the game situation (McLean et al., 1999). Sidestep cutting angles were required to be 35°–55° from the original direction of motion, again reflecting values typically demonstrated in the game situation, and adopted previously (McLean et al., 2004a,b). Angles were measured from the center of the force plate and the corresponding line was marked (using tape) so that it could be clearly seen by the subjects (Fig. 1). Specifically, subjects were required to land and sidestep cut off the right leg, such that the cutting action moved the subject forward and to the left of the force plate at the appropriate angle (McLean et al., 1999, 2004a,b) (see Fig. 1). Kinematic data were obtained from the 3D coordinates of skin-mounted markers secured to various anatomical locations (Fig. 2), recorded via six electronically shuttered high-speed video cameras at 240 fps and Eva 6.0 tracking software (Motion Analysis Corp., Santa Rosa, CA, USA). A standing trial was first collected with all joints in the neutral position, following which, the forehead, left and right anterior superior iliac spine (ASIS), medial femoral condyle and medial and lateral malleoli markers were removed prior to the motion trials. Synchronized 3D GRF data were collected during each trial at 1000 Hz via an AMTI force plate (Model OR6-5, Serial # 4068, Watertown, MA, USA).

Table 1
Mean (SD) subject characteristics by gender

Characteristic	Gender	
	Male ($n = 10$)	Female ($n = 10$)
Age (years)	20.2 (1.9)	21.1 (3.0)
Experience (years) ^a	10.2 (5.1)	10.5 (4.8)
Height (cm)	184.7 (8.0)	176.0 (11.1)
Weight (kg)	81.9 (9.8)	76.1 (12.4)
Femur length (cm)	49.6 (4.5)	44.5 (3.8)
Tibia length (cm)	41.7 (3.6)	39.2 (3.6)
ASIS width (cm)	28.7 (5.0)	31.7 (5.0)
Femoral condyle width (cm)	10.5 (0.9)	10.9 (0.6)

^a Experience was denoted by the number of years participating in organized sporting (basketball) activity.

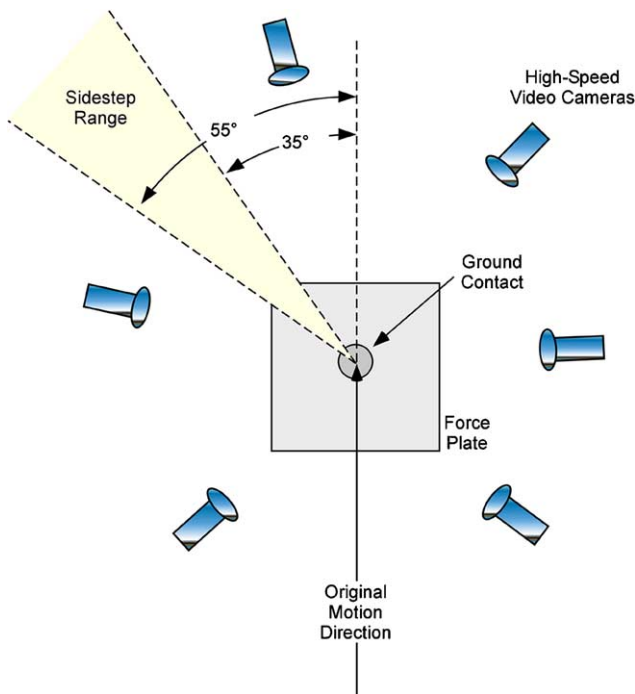


Fig. 1. Successful sidestep execution was based on the movement occurring on a force plate, within the field of view of a high-speed video system and within a prescribed cutting range.

2.2. Model development and validation

A detailed description of the model development and validation procedures has been presented previously (McLean et al., 2003). Briefly, standing trial data obtained for each subject were used with Mocap Solver 6.17 (Motion Analysis Corp., Santa Rosa, CA, USA) to define a kinematic model comprising five skeletal segments (foot, talus, shank and thigh of the support limb, and the pelvis) and 12 degrees of freedom (DoF). The pelvis had six DoF relative to the global (lab) coordinate system, with the hip, knee and ankle joints defined locally and assigned three, one and two rotational DoF respectively (McLean et al., 2003) (Fig. 3). The 3D marker trajectories recorded during the 10 sidestep cutting

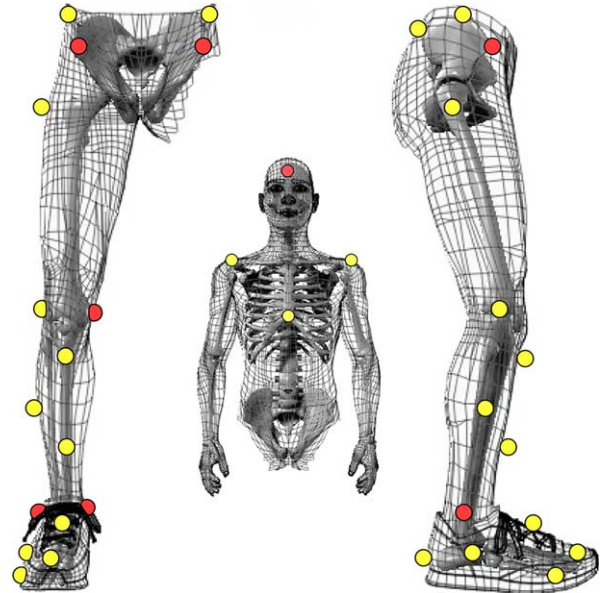


Fig. 2. Marker locations used to define a kinematic model comprised of 5 skeletal segments. The forehead, left and right ASIS, medial femoral condyle and lateral and medial malleoli markers (red) were removed for the recording of movement trials. (For interpretation of the references in colour in this figure legend, the reader is referred to the web version of this article.)

trials for each subject were then processed by the Mocap Solver software to solve for the twelve DoF of the skeleton model at each time frame (0–200 ms).

A forward dynamic 3D rigid body model of the trunk and lower extremity was developed for each subject consisting of the skeletal model described above, with wobbling masses added to the pelvis and thigh segments. The mass attached to the pelvis represented all body segments that were not modeled, including the non-support limb, arms and head (McLean et al., 2003). Contact between the foot segment and ground was modeled using 91 discrete viscoelastic elements, with each element attached in 3D locations describing the exterior shoe surface. Model inertial characteristics were based on anthropometric data obtained for each subject (de Leva, 1996). Equations of motion for each model were produced by SD/FAST (Parametric Technology Corp., Needham, MA, USA).

Thirty-one muscles were attached to the skeleton (Delp et al., 1990), which were categorized into 12 functionally discrete groups. A three-element Hill model was used to model muscle-tendon dynamics as described previously (McLean et al., 2003), but with all model parameters taken from SIMM (Software for Interactive Musculoskeletal Modeling) (Musculographics, Chicago, IL, USA). For computational efficiency, the 3D muscle path models from SIMM were converted into a multivariate polynomial for musculotendon length as a function of the joint angles $q_1 \dots q_M$ between origin and insertion (Dhillon and van den Bogert, in press):

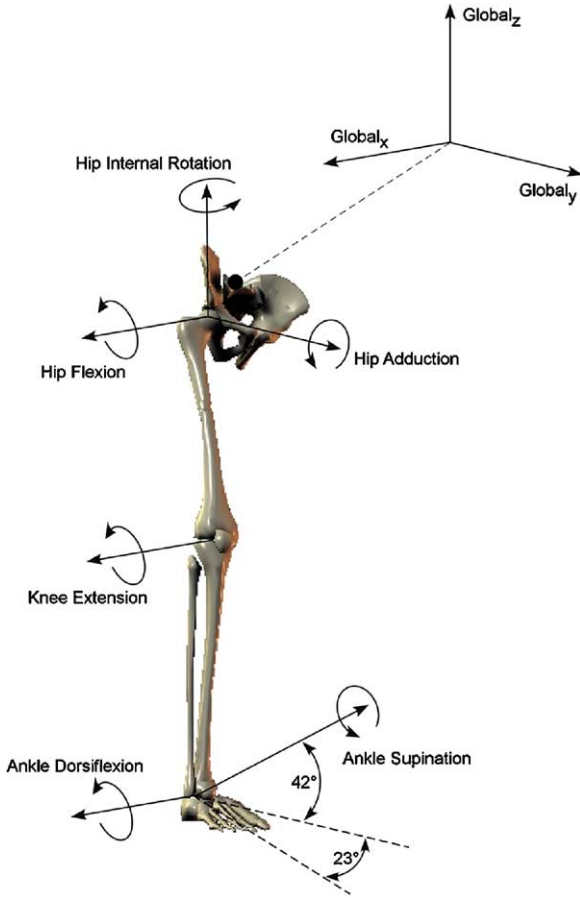


Fig. 3. For the kinematic model, Pelvis (body) motion was described with respect to the Global (lab) coordinate system via three translational and three rotational degrees of freedom. The hip, knee and ankle joints were defined locally and assigned three, one and two rotational DoF respectively (McLean et al., 2003).

$$L_m = \sum_{i=1}^N A_i \prod_{j=1}^M q_j^{E_{ij}} \quad (1)$$

The model parameters (N polynomial coefficients A and NM integer exponents $E \geq 0$) were found by stepwise polynomial regression on muscle moment arm data generated by SIMM at various combinations of joint angles. During the movement simulations, muscle moment arms were obtained by partial differentiation (An et al., 1984) of Eq. (1).

Neural stimulation inputs for each muscle group were modeled as a piecewise linear function of time, with five parameters: the stimulation value at times 0, 50, 100, 150, and 200 ms after heel strike. Body segment positions and velocities quantified at heel strike for each subject, were averaged over the 10 sidestep cutting trials and used as initial conditions for the forward dynamic simulations. An ensemble average (SD) was calculated across the 10 trials for the nine rotations and three GRF's. Muscle stimulation patterns were optimized via a simulated annealing algorithm (Goffe et al., 1994) to mini-

mize the difference between simulated and these baseline data. Root-mean-square (RMS) fit errors and RMS prediction errors were quantified as described in McLean et al. (2003) for each of the twelve variables of interest, and were used to assess model validity.

2.3. Extraction of resultant knee joint loads

For each optimized system, the resultant anterior-posterior joint reaction force (FR_{AP}), varus-valgus (adduction-abduction) (M_{VV}) and internal-external (M_{IE}) reaction moments, with respect to the tibial anatomical reference frame, were obtained from the dynamic equations of motion at 1ms intervals. The relative contributions of the quadriceps and hamstring muscle forces to the anterior-posterior joint load were calculated using equations for tendon orientation as a function of knee flexion angle (Herzog and Read, 1993). These contributions were added to the resultant load FR_{AP} to obtain an estimate of the anterior drawer force (FD_{AP}). The knee joint coordinate system orientation was such that external anterior drawer force, the anterior component of the quadriceps force, and varus and internal rotation loads applied to the joint were all defined as positive.

2.4. Neuromuscular control effects on knee loading

Monte Carlo simulations ($n=5000$) were performed with each model to determine the effects of variability in NMC on peak anterior drawer force (FD_{Ant}), valgus moment (M_{Val}) and internal rotation moment (M_{Int}) data during the first 200 ms of sidestep stance. Specifically, for each of the 5000 simulations, random numbers were added to the initial body segment and angular positions and linear and angular velocities. These numbers were generated from a Gaussian distribution with zero mean and the standard deviation in each movement variable calculated across the 10 sidestep cutting trials. Optimized stimulation parameters for the knee extensors (rectis femoris and vasti group) and knee flexors (hamstring group) were each multiplied by a separate Gaussian random number with a mean of one and a standard deviation of one. Muscle stimulation levels were limited to values between zero and one as per the model setup (McLean et al., 2003).

2.5. Data analyses

Peak stance (0–200 ms) phase values for FD_{Ant} , M_{Val} , M_{Int} and anterior joint reaction force (FR_{Ant}) obtained from each optimized simulation were submitted to a one-way ANOVA to determine for the main effect of gender. With Bonferroni correction, an alpha level of 0.013 was required for statistical significance. Effect size was also determined for each comparison according to

Cohen (1988), where by definition, large, medium and small effect sizes were denoted by values greater than 0.8, 0.5 and 0.2 respectively. Peak FD_{Ant} , M_{Val} and M_{Int} data were also recorded for each of the 5000 randomly perturbed simulations in each subject. The potential for sagittal plane loading as an ACL injury mechanism was quantified as the number of simulations where peak FD_{Ant} exceeded 2000 N. This value was chosen based on ultimate failure loads reported previously for the ACL (Woo et al., 1991).

3. Results

After optimization of the subject-specific movement simulations, the fit and prediction errors were similar for male and female models (Table 2). For each of the 12 optimized model variables, the mean difference between measured and simulated data was less than two standard deviations. In fact, excluding the GRF data, mean differences were less than one standard deviation. RMS prediction errors were typically between 1.5 and 3.5 (see Table 2). However, mean prediction errors were larger in both male and female models for pelvis somersault angle and ankle pronation–supination angle. The mean (SD) optimized muscle activation parameters ($n=5$) for the rectus femoris, and vastus and hamstring muscle groups were consistent between individuals and genders (Fig. 4).

Mean external load patterns obtained from the optimized simulations of sidestep stance, were similar for male and female models (Fig. 5). Gender comparisons of peak joint loads in the optimized movement simulations revealed that males had larger M_{Int} during the stance phase of the sidestep cut than females (Table 3). A large effect size was also observed for this comparison. A medium effect size was also calculated for peak

M_{Val} comparisons, with females demonstrating noticeable increases in mean values compared to males.

Random perturbations in initial body and segment positions and velocities produced noticeable increases in peak FD_{Ant} , M_{Val} and M_{Int} values for both male and female model simulations compared to mean optimized values (Fig. 6). Despite these increases however, peak FD_{Ant} measures never exceeded 2000 N in any model. Hence, no ACL injuries were reported for the sagittal plane loading mechanism.

4. Discussion

This study examined the potential for sagittal plane biomechanics associated with sidestep cutting to be an isolated mechanism of ACL injury. The extent to which this relationship may be dependent on gender was also evaluated. Testing these postulates necessarily required knee joint loading associated with actual injury-causing events to be examined. Forward dynamic simulations of sidestep cutting movements, such as that presented here appear therefore to provide the greatest potential for successful elucidation.

4.1. Model validity

Mean validation (RMS/Fit) errors for both male and female models (see Table 1) were similar to those reported previously for a single subject (McLean et al., 2003). Specifically, all simulated variables fell within the pre-defined criteria of two standard deviations from the measured data. The lower limb joint kinematics quantified during sidestep cutting for each subject were also consistent with those reported previously (Colby et al., 2000; McLean et al., 1999, 2004a; Neptune et al., 1999). Based on these results, optimized models

Table 2
Mean (SD) validity measures for optimized model simulations of sidestepping as a function of gender

Variable	RMSFit/SD		RMSPred/SD	
	Male	Female	Male	Female
Medio-lateral force (F_x)	1.22 (0.50)	1.31 (0.66)	1.88 (0.44)	2.15 (0.62)
Anterior–posterior Force (F_y)	1.54 (0.76)	1.80 (0.54)	2.54 (0.90)	2.65 (0.72)
Vertical force (F_z)	1.32 (0.49)	1.45 (0.68)	2.45 (0.61)	2.43 (0.84)
Somersault (R_x)	0.75 (0.41)	0.92 (0.57)	3.76 (1.40)	4.28 (2.60)
Tilt (R_y)	0.71 (0.31)	0.85 (0.37)	3.05 (1.06)	3.21 (1.53)
Twist (R_z)	0.71 (0.66)	0.89 (0.73)	2.00 (0.58)	2.30 (1.22)
Hip flexion–extension (H_x)	0.60 (0.19)	0.80 (0.43)	2.65 (1.06)	3.36 (1.85)
Hip abduction–adduction (H_y)	0.61 (0.37)	0.63 (0.21)	3.02 (1.18)	2.25 (0.61)
Hip axial rotation (H_z)	0.83 (0.57)	0.64 (0.28)	2.52 (1.05)	2.56 (1.33)
Knee flexion–extension (K_x)	0.60 (0.24)	0.69 (0.30)	2.77 (1.28)	3.09 (1.06)
Ankle planter–dorsi flexion (A_x)	0.93 (0.43)	0.86 (0.33)	3.38 (2.01)	3.33 (1.36)
Ankle pronation–supination (A_y)	0.54 (0.23)	0.65 (0.33)	3.92 (1.25)	5.24 (2.28)

RMS fit error corresponds to the average difference in terms of SD's between simulated and measured data. RMS prediction error is the ratio of the mean RMS difference between the 10 sets (trials) of measured and simulated data, to the mean measured inter-trial variability over 200 ms.

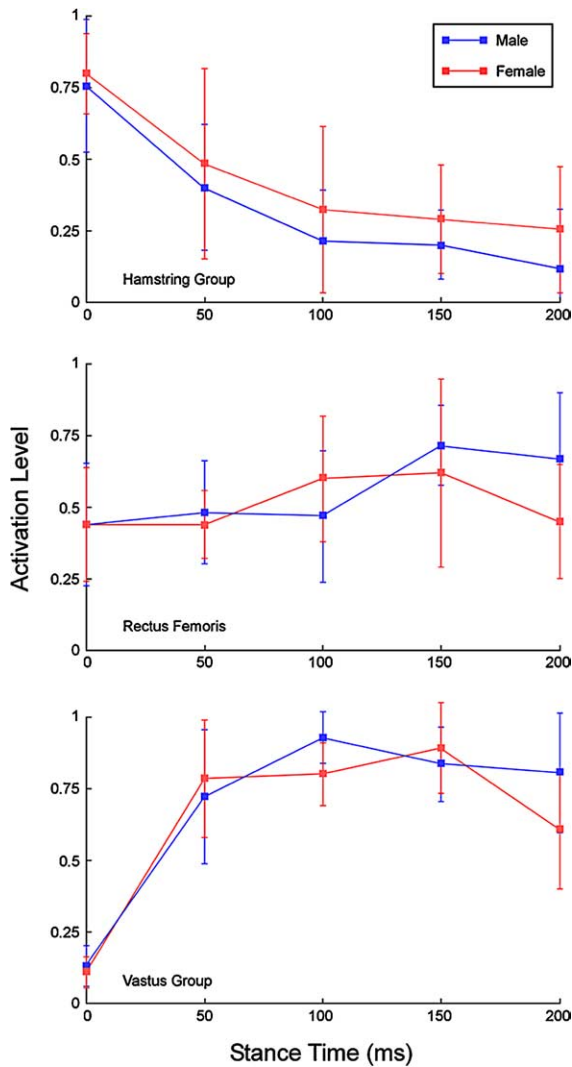


Fig. 4. Gender comparisons of mean (\pm SD) hamstring, rectis femoris and vastus muscle activation patterns obtained for the first 200 ms of sidestep stance in optimized model simulations.

were deemed to successfully simulate realistic sidestep cutting maneuvers in each subject.

Poor or abnormal NMC during sidestep cutting execution has become increasingly viewed as a major contributor to ACL injury risk (Boden et al., 2000; Griffin et al., 2000; Lephart et al., 2002). Therefore, the ability of models to predict the consequences of perturbed NMC was viewed to be crucial. Mean normalized RMS prediction errors for male and female models (Table 1) were consistent with those presented previously (McLean et al., 2003). In some instances, such as for whole body rotations, noticeable improvements were seen, possibly due to the use of a more detailed musculoskeletal model. As with the original model however, prediction errors for ankle supination-pronation were quite large, which may be caused by our relatively simple foot model with two degrees of freedom at the ankle and no intrinsic foot joints. A sensitivity analysis was

conducted to assess the impact of this potential limitation on current results. We found that knee joint loading was not particularly sensitive to changes in ankle supination-pronation patterns. However, incorporation of a better representation of foot and ankle should be considered in future model developments.

4.2. External knee loads for optimized simulations

After the successful optimization and validation of subject-specific sidestep cutting simulations, 3D knee joint loads could be extracted from each model with confidence. Three of the four loading variables (FR_{AP} , M_{VV} and M_{IE}) were obtained directly from the SD/FAST multibody software as resultant external joint loads. These variables are essentially the same as those that would be obtained using a standard inverse dynamics approach using the same kinematic and GRF data. Generating these data via a forward dynamic optimization however, allowed us to also predict how they are affected by NMC and produce potential injury scenarios.

Mean stance phase patterns for M_{VV} and M_{IE} were consistent with our original findings (McLean et al., 2003) and with those presented previously for sidestep cutting (Besier et al., 2001). However, the peak magnitudes were noticeably larger than those reported by (Besier et al., 2001). Differences in experimental methodology, particularly in terms of the cutting angles and speeds adopted in each study may explain the concomitant differences in load magnitudes. Differences in subject skill or experience level between the two studies may also be an important contributing factor, particularly in terms of how aggressively the maneuvers were performed, ultimately manifesting in knee loading parameters (McLean et al., 2004a,b).

Female models had higher valgus and decreased internal rotation torques than males. Corresponding experimental data comparing male and female 3D external joint loads during sidestep cutting execution does not exist. However, the above differences are consistent with those observed previously for gender comparisons of lower limb joint kinematics during sidestep cutting (Malinzak et al., 2001; McLean et al., 2004a,b) and jump landing (Ford et al., 2003) tasks. Such differences are suggested to stem from concomitant gender-based differences in lower limb anatomy (McLean et al., 1999) and NMC during movement execution (Boden et al., 2000; Griffin et al., 2000; McLean et al., 2004a). These assertions appear substantiated considering that in the current case, lower limb alignment and initial contact (NMC) conditions for each model were subject specific. The impact of these differences in terms of ACL injury risk will be discussed later in more detail.

Mean patterns for FR_{AP} were also consistent with those reported for our original model (McLean et al., 2003). Further, differences were not observed in these

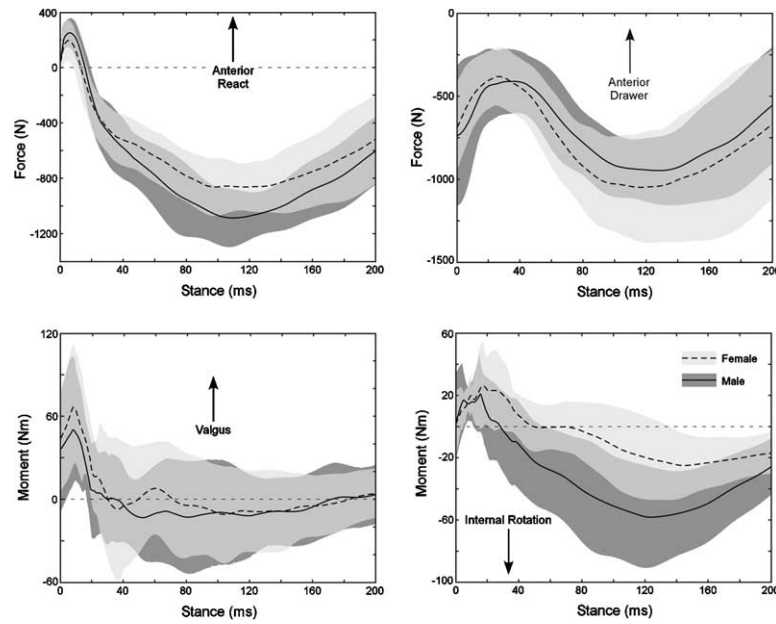


Fig. 5. Comparisons of mean (\pm SD) male and female knee joint forces (FR_{AP} and FD_{AP}) and moment (M_{VV} and M_{IE}) quantified over the first 200 ms of sidestep stance in optimized model simulations.

Table 3

Effect of gender on mean (SD) peak joint loads estimated during the stance phase (0–200 ms) of optimized sidestep cutting simulations

Dependent measure	Male	Female	Effect size r
FR_A (N)	284.2 (126.3)	240.9 (136.9)	0.33
FD_A (N)	–482.4 (157.0)	–472.5 (200.7)	0.05
M_{Val} (Nm)	69.8 (23.0)	83.5 (24.2)	0.58
M_{Int} (Nm) ^a	71.1 (24.5)	33.4 (14.7)	1.87

FR_A = external joint reaction force; FD_A = anterior drawer force; M_{Val} = valgus moment; M_{Int} = internal rotation moment.

^a Denotes statistically significant difference between genders ($p < 0.013$).

data between male and female models. Specifically, a net anterior knee joint reaction force was evident during the initial weight-acceptance phase of the sidestep cut. A posterior knee joint force was then observed for the remainder of stance (see Fig. 4). This joint loading pattern is likely dominated by the large posteriorly directed force acting on the tibia during stance, which stems from the posterior external GRF during deceleration (McLean et al., 2003). The large magnitude of the posterior joint reaction force during sidestep stance suggests that the impact of this pre-mentioned mechanism may be more important than had been considered previously in theories pertaining to ligament injury. This concept will be expanded upon further when injury potential is discussed.

During sidestep cutting, the external joint reaction loads are counteracted to a large extent by the force action of the surrounding musculature, with the net resultant loads being taken up by the passive joint structures (Lloyd and Besier, 2003). Evaluating the potential for

injury in these structures therefore, necessarily requires the loads they experience to be known. We chose the net resultant sagittal plane load, more specifically the anterior–posterior drawer force (FD_{AP}), to denote ACL loading during the simulated sidestep cutting tasks. As noted earlier, net drawer force was obtained from the summation of the anterior–posterior resultant joint reaction force and the anterior–posterior force actions of the quadriceps and hamstring muscles. Similar methods have been used previously to provide estimates of ACL loading during skiing (Gerritsen et al., 1996), open and closed chain knee extension (Escamilla et al., 2001) and sidestep cutting (Simonsen et al., 2000). Considering that the ACL is the primary restraint to loading (anterior) in this plane (Butler et al., 1988), this representation appears feasible.

Mean estimates of peak anterior drawer force were never found to be positive in either male or female models. This result implies that the ACL is not significantly loaded via the sagittal plane mechanism during typical sidestepping movements. This observation is in direct contrast to the work of (Simonsen et al., 2000), where a mean ACL force (anteriorly directed shear force) of 520 ± 68 N was estimated for the stance phase of sidestep cutting tasks. The difference in load response may be due to the different methods for estimating muscle co-contraction. Similar to the current case however, (Escamilla et al., 2001) did not observe ACL forces during simulated leg press and squatting exercises. (Cerulli et al., 2003) has measured in vivo ACL strain in a single male subject performing a rapid deceleration task, and found peak strains of $5.47 \pm 0.28\%$, corresponding to the peak in vertical GRF. External knee joint loads were

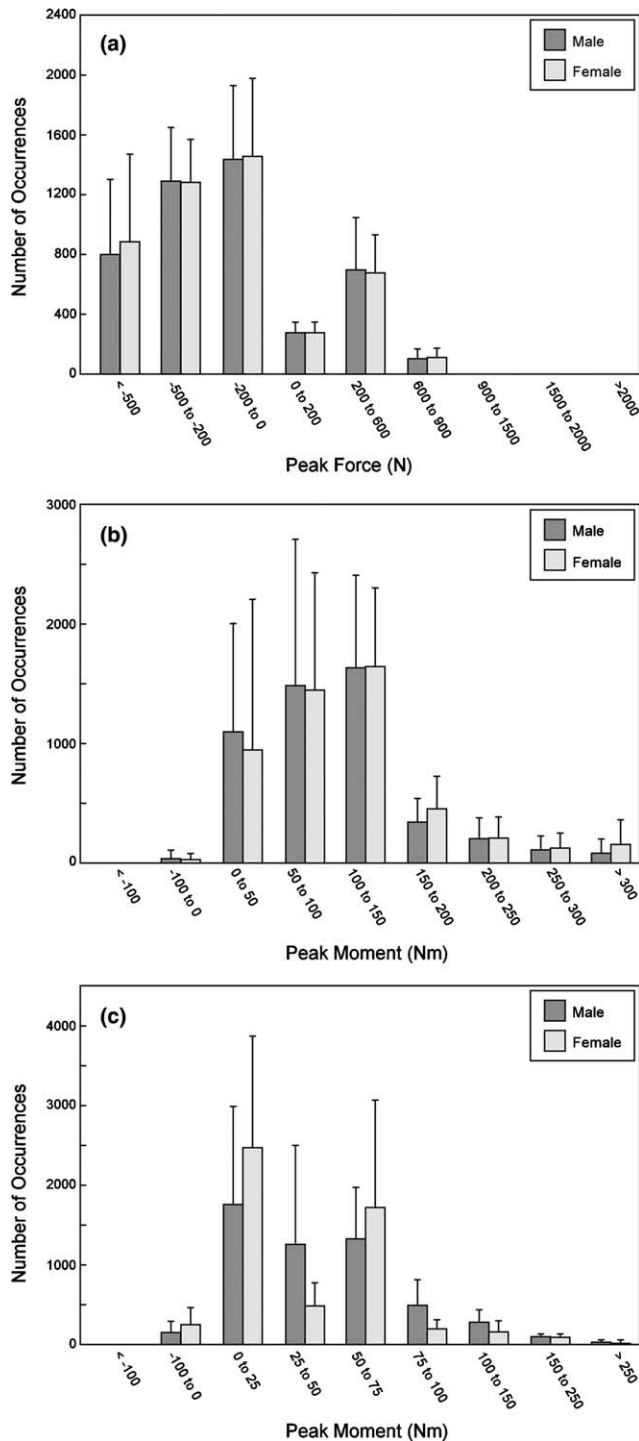


Fig. 6. Effect of initial contact NMC perturbations ($n = 5000$) on mean resultant anterior drawer force (a) and valgus (b) and internal rotation (c) moments quantified in male and female sidestep models. ACL injury was deemed to occur when anterior drawer force exceeded 2000 N (Woo et al., 1991).

not measured however, and hence, the extent to which the ACL strain response was a direct result of the sagittal plane load mechanism is unknown. It is possible for example, that ACL strain resulted from simultaneous out-of-plane joint loading.

It is possible that for our optimized models, the anterior drawer force, and hence estimates of ACL load were underestimated. During early stance for example, modeled hamstring activations were high while at the same time quadriceps activations were low (Fig. 4). Subsequently, FD_{AP} was dominated by the action of the hamstrings, resulting in a large posteriorly directed shear load (see Fig. 5). Previous EMG studies of sidestep cutting however, have reported hamstring activation to be relatively low at heel contact and remain that way throughout the entire deceleration phase (Colby et al., 2000; Neptune et al., 1999; Simonsen et al., 2000). It may be therefore that in our model, hamstring contributions to ACL loading were greater than in reality. Large hamstring forces may have been required to control upper body motion in the models, as no muscles or joints were included for that purpose (McLean et al., 2003). It should be noted however, that more recent studies have observed hamstring activation patterns that are consistent with our model outputs, for both rapid deceleration (Cowling and Steele, 2001) and sidestep cutting tasks (Besier et al., 2003). In these cases, hamstring activations were viewed as a pre-planned strategy to counter ACL loading upon landing. Further research characterizing the lower limb EMG response during sidestep cutting in both males and females appears necessary.

4.3. Potential for ACL injury in the sagittal plane

While it is possible that ACL forces were underestimated in optimized models, this was not the case for ACL injury simulations. Specifically, applying random perturbations of up to 100% to optimized muscle activation patterns over a series ($n = 5000$) of (Monte Carlo) simulations necessarily resulted in instances where hamstring force remained at zero, while conversely, quadriceps forces were doubled. Situations where this occurred in combination with the knee joint at or near full extension represented a “worst-case-scenario” in terms of sagittal plane contributions to ACL injury risk (Colby et al., 2000; Durselen et al., 1995; Pandy and Shelbourne, 1997). Thus, all injury possibilities were effectively explored via this method.

Random perturbations in initial body and segment kinematics, and in muscle activation patterns, representing realistic variations in NMC, produced considerable increases in peak anterior drawer during sidestep stance. Despite these increases however, forces were never large enough to produce ACL injury, being well below the pre-determined injury threshold of 2000 N (Woo et al., 1991). Specifically, peak anterior drawer forces never exceeded 900 N regardless of the applied neuromuscular perturbations. These observations are consistent with our original findings, where a peak anterior drawer force of 872 N was observed over 100,000 Monte Carlo

simulations in a single subject-specific model (McLean et al., 2003).

The fact that the sagittal plane loading mechanism did not in isolation cause ACL injury can be explained as follows. As noted above, large quadriceps forces applied at or near full knee extension, in conjunction with minimal hamstring activity offers the greatest potential for a sagittal plane injury mechanism. In this position, the angle between the patellar tendon and tibial long axis is such that large anterior shear loads are possible (Pandy and Shelbourne, 1997). With the knee in this position however, muscle fibers in the quadriceps are shortened such that their maximum force production is significantly reduced (Delp et al., 1990). Conversely, if the knee is flexed at contact such that the quadriceps produces a large force, the patellar tendon will simultaneously be more parallel to the tibial axis, effectively reducing the magnitude of quadriceps-induced anterior shear (Herzog and Read, 1993; Pandy and Shelbourne, 1997). The interaction between the quadriceps and the anterior-posterior GRF's during sidestep cutting may also contribute to the apparent ceiling on maximal sagittal plane loading. As noted earlier, the rapid deceleration associated with the stance-phase of the sidestep creates a posteriorly directed external force vector at the shoe ground interface, which is transferred to the tibia, and helps protect the ACL. Due to the moment balance in the sagittal plane, increased quadriceps force will necessarily be associated with an increased posterior GRF. The net change in ACL loading via the action of the quadriceps in this instance will therefore be significantly reduced.

Based on current observations, it appears that other loading mechanisms apart from that linked to the sagittal plane are necessary during sidestep cutting to produce an ACL injury. Previous research has shown that valgus and internal rotation knee loads, both in isolation and in combination, have a significant impact on ACL loading (Kanamori et al., 2000; Markolf et al., 1995). (Seering et al., 1980) have also shown that ligament damage occurred in cadaveric knee joints within 125–210 Nm of valgus torque or 35–80 Nm of internal rotation torque. Significant out-of-plane loading was evident for the optimized sidestep cutting models and Monte Carlo simulations produced peak valgus and internal torques well above these injury ranges. Thus, out-of-plane loads large enough to injure the ACL may be possible during sidestep execution. Furthermore, it appears that knee valgus loading is the 3D knee loading variable that is most sensitive to changes in NMC during sidestep cutting. This observation is consistent with our original findings (McLean et al., 2003). The fact that models based on female data produced more instances of hazardous valgus loading (see Fig. 5) also suggests that this variable may be an important contributor to the gender disparity observed in the risk of ACL

injury. Recent kinematic (Ford et al., 2003; Malinzak et al., 2001; McLean et al., 1999, 2004a) and prospective (Hewett et al., 2004) studies similarly propose knee valgus and valgus loading to be key predictors of ACL injury in females. The means by which lower limb NMC parameters may manifest in terms of valgus loading during movements such as sidestep cutting requires further investigation.

While the sagittal plane loading mechanism does not appear able to in isolation injure the ACL during sidestep cutting, it may still contribute indirectly to injury risk via its ability to limit and/or control out-of-plane loads such as knee valgus. (Besier et al., 2003), has shown that sagittal plane muscle activation strategies (quadriceps and hamstrings) can influence the ability to stabilize the knee joint in varus-valgus and internal-external rotations. There may be instances therefore, where the combined force action of these muscles cannot effectively counter the associated valgus loading, thus subjecting the ACL to larger and potentially hazardous loads. We have recently shown that apart from demonstrating increased knee valgus, females also land in a more (hip and knee) extended position during sidestep cutting and jump landing tasks compared to males (McLean et al., 2004a,b). It may be that this landing posture does not afford optimal force control of the sagittal plane muscle groups in terms of valgus loading. It has also been suggested however, that these postures may represent pre-planned strategies that attempt to minimize the potential for extreme out-of-plane loading scenarios (Besier et al., 2003; Cowling and Steele, 2001). Further work appears necessary to determine whether a causal link exists between sagittal plane biomechanics and valgus loading during sports movements such as sidestep cutting, and thus, how this mechanism may be altered/trained to reduce the likelihood of ACL injury.

4.4. Limitations

One important model simplification that may have affected results was that internal-external rotation of tibia with respect to the femur was not included in the model. With all internal-external rotations transferred to the hip joint, it was therefore possible that out-of-plane knee loads were overestimated in the models. It should be noted however, that this potential limitation would not have affected sagittal plane load calculations. A sensitivity analysis has also shown that modeling the knee joint in this fashion has only a minor impact on model performance (McLean et al., 2003). An additional consideration was that accurate measurement of internal-external knee rotation in the subject would be needed, which is almost impossible due to skin marker artifacts (Reinschmidt et al., 1997) and might have introduced additional error into the movement simulations.

Descriptions of ACL loading, and hence, predictions of injury potential were based on peak sagittal load only. While it is known that non-sagittal moments contribute to ACL loading, the influence of combined knee loading states on resultant ACL load has been quantified for relatively low loading states only (Kanamori et al., 2000; Markolf et al., 1995). Such loads however, are not representative of the extreme joint loading postures associated with sports movements such as sidestep cutting. A quantitative understating of ACL loading during these movements is therefore imperative, if injury mechanisms and the extent to which they may be gender specific are to be identified in the future.

5. Conclusions

- (1) During normal sidestep cutting movements, the sagittal plane loading mechanism does not generate ACL loading.
- (2) During normal sidestep cutting movements, knee valgus moment was higher in females, and peak internal rotation moment was higher in males.
- (3) Sagittal plane forces applied to the knee joint during sidestep cutting as a result of realistic neuromuscular control perturbations cannot cause ACL injury.
- (4) Neuromuscular control perturbations can cause knee valgus loads that are large enough to injure the ACL.
- (5) There is a need to quantify ACL loading for extreme 3D knee joint loading scenarios typical of hazardous sporting postures.

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